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An active cardiac stabilizer based on gyroscopic effect

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***Abstract*— Active cardiac stabilization has a role to play in the development of minimally invasive techniques for beating heart surgery. We propose here a new active cardiac stabilization device based on gyroscopic actuation. This system allows to compensate for heart motion in high frequencies and is fully independant and pluggable on conventional stabilizers. The mechanical model and design are described. The system is controlled thanks to static state feedback, taking into account gyroscope specificities. Experiments results are presented. They highlight the effectiveness of this solution with a 48% reduction of the RMS excursion.**

I. Introduction

A general trend in the development of new surgical techniques is the reduction of the invasiveness. Laparoscopic surgery is now widely used in particular for gastrointestinal, gynecological and urological surgery. Extending these techniques to other fields is a crucial challenge since their benefits are numerous: e.g. it reduces infectious hazard, scars size, hospital stay and patient recovery time. However, in some cases, as for cardiac surgery, the use of minimally invasive techniques is not applicable easily as pointed out in [1], even though it could heavily improve surgery quality. Indeed a common operation like coronary artery bypass implies heavy invasive steps such as sternum dissection, rib cage opening and extra corporeal circulation which are the main complication causes and could be avoided using minimally invasive beating heart surgery. Nevertheless dealing with the heart motion is the main hindrance to the use of such techniques.

However, some beating heart coronary bypasses have been performed by open and laparoscopic ways as in [2]. In these cases, the cardiac motion problem has been compensated for using passive stabilizers which are constituted by a stiff rod maintaining the area of interest on the myocardium thanks to two pressure or suction fingers. But with these stabilizers the residual motion due to the device flexibility is not negligible as highlighted in [3]. An alternative solution based on active compensation was developed by Bachta *et al.* [4], demonstrating the efficiency of active stabilization on motion reduction. This solution is particularly efficient for low frequencies but suffers from a limited bandwidth because of proximal actuation. In addition it requires the use of specific instruments. The solution presented herein is based on an alternative actuation method and carries out active stabilization for higher frequencies which are the most problematic for surgeons. Moreover it is fully independant and is designed to be pluggable to existing commercial instruments while satisfying the sterilizabitity constraints.

For this purpose, we chose to take advantage of the gyroscopic effect. This mechanical property is known to allow the generation of torques without the need to be linked to the ground since it is based on inertial effects. Various applications using this concept can be found in the literature: stabilization systems for ships [5], attitude control in zero gravity environments for satellites [6] and submarine robots [7], antisismic building stabilization [8] and non-grounded haptic interface [9] for instance. This principle is usually used for rather large and heavy structures, however we propose to use it at a smaller scale, within the framework of cardiac stabilization for beating heart surgery. Moreover, since the system presented here is independent from the surgical instrument, the concept can be extended to other applications needing structure stabilization in similar frequency domain.

This paper deals with the design of this system, from concept to first experiments. First, mechanical aspects are presented, including equations of the model and design issues. Then the control law we chose and which meets the specificities of the gyroscope is explained. Finally the first experimental results are presented.

II. System design

A. Problem specification

Since the system should compensate for heart actions, the corresponding forces have to be identified. For this purpose, assessment of forces applied on a passive stabilizer distal end were performed *in vivo* in real operation conditions on an anesthezied pig [4]. The data revealed that force and displacements are prominent along the vertical direction confirming the results from [10]. Hence we chose to compensate actively for residual motions along this direction. Frequency-domain data analysis highlighted several distinct actions: a constant value due to initial constraint needed to maintain the heart that prevents stabilizer unsticking, breathing components with a fundamental frequency of 0.25 Hz, and heart beating components starting at 1.5 Hz. In the sequel we will focus on the compensation of cardiac component only, which is the core of the problem. Indeed the breathing motion is not a problem specific to cardiac surgery and surgeons are able to deal with in most cases, so compensation is not indispensable for low frequencies.

B. Mechanical model

The rigid-body model of the gyroscopic compensator, which is depicted in Fig. 1, is composed of an inertia wheel

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④ rotating at a constant high speed Ω with respect to the joint q_4 thanks to a first actuator. This constitutes the gyroscope. This wheel is attached to a gimbal ③ which rotation with respect to the joint q_3 can be controlled with a second actuator. The whole system is attached to the rod of the passive stabilizer ② so that the gyroscope axis is aligned with A at its nominal position. The passive stabilizer is connected to the ground by a massless part ① using two revolute joints q_1 and q_2. Those latter joints allow to model the stabilizer attachment flexibility which stiffness and damping factor are respectively denoted k and f.

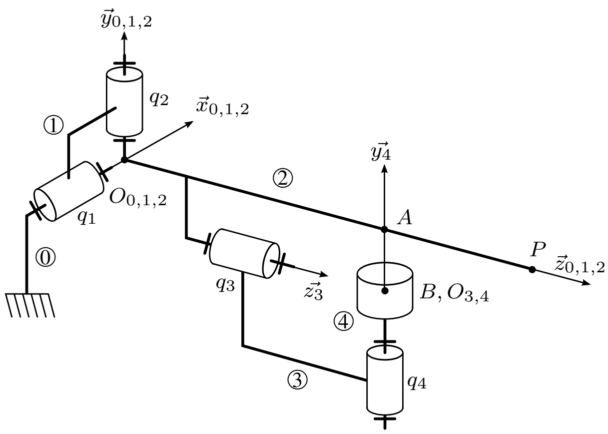


Fig. 1. System model in its nominal configuration.

We introduce the following parameters for the system modeling. The total length of the stabilizer is L and its distal-end point is P, L_A denotes the distance OA and L_B the distance AB. J_2 denotes the stabilizer moment of inertia with respect to the axes $\vec{x_2}$ and $\vec{y_2}$. The mass of the gyroscope is m_4 and its moment of inertia with respect to its revolution axis $\vec{y_4}$ is B_4. The forces F_x and F_y denote the projection on the horizontal and vertical axis of the cardiac force applied at point P. The Lagrange equations, linearized around the position $(q_1, q_2) = (0, 0)$ and simplified to eliminate negligible inertial effects of the gyroscope, are the following:

$$\begin{aligned}\mathcal{L}_{q_1} : \left[J_2 + m_4\left(L_A^2 + L_B^2\right)\right]\ddot{q}_1 + f\dot{q}_1 + kq_1 \\ = \cos(q_3)T_G - F_yL\end{aligned} \tag{1}$$

$$\begin{aligned}\mathcal{L}_{q_2} : \left(J_2 + m_4L_A^2\right)\ddot{q}_2 + f\dot{q}_2 + kq_2 \\ = \sin(q_3)T_G + F_xL\end{aligned} \tag{2}$$

$$T_G = B_4\dot{q}_3\Omega \tag{3}$$

One can note in (3) that the gyroscopic torque T_G is proportional to the gimbal speed, the gyroscope speed and its moment of inertia. Since Ω and B_4 are constant it is possible to impose in real time the torque required to compensate for heart motion by controlling the gimbal speed $\dot{q}_3$. Finally this system generates a torque proportional to the speed input. We can also note that the torque direction depends on the gimbal angle value q_3. Hence if we want to generate a torque to compensate for the motion along $\vec{y}$ only, the gimbal should remain in its nominal position.

C. Design issues

For the design, the passive stabilizer is modeled by a stainless steel rod with a diameter of 10 mm and a length of 300 mm which corresponds to commercial stabilizers in terms of dimensions and rigidity.

The compensator should be as compact and light as possible and able to produce a high gyroscopic effect. For this purpose we refer to equation (3). The torque value is determinated from the cardiac force, and $\dot{q_3}$ from the minimum compensation frequency and the maximum gimbal acceptable angle. Hence the design choice concerns B_4 and Ω. We finally chose to maximize the gyroscope spin rate rather than its inertia which would induce higher weight. Another important constraint on the design is to provide a motion compensation device compatible with existing passive stabilizers and independant to allow separate sterilization. Most parts of the system are made of aluminum to improve lightness except the gyroscope which is made of steel to increase its inertia. The gyroscope and gimbal axis are both guided thanks to stainless steel bearings. Mechanical design was validated after successful static and modal finite element analysis using a worst case scenario. A picture of the system designed to comply with the foregoing requirements is presented in Fig. 2.

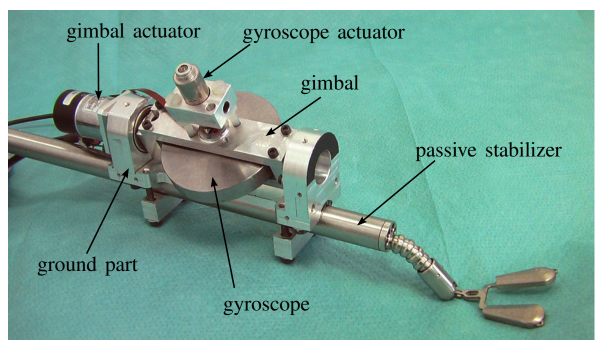


Fig. 2. System overview, here mounted on a passive stabilizer from Medtronic.

Finally the designed system is 130 mm long and wheights 390 g including actuators. Thus the compactness is clearly improved compared to previous solutions.

III. Control strategy

A. Model for the synthesis

The aim is to control the gimbal in order to stabilize the beam position along a vertical axis. As can be seen in (3), the torque provided by the gyroscope in the $\vec{x}$ direction is proportional to $\dot{q}_3$ and $\cos(q_3)$ since Ω and B_4 are constant. So it can be controlled using the gimbal speed as an input. Since the gimbal position is available, we consider in the sequel that the nonlinear $\cos(q_3)$ term is linearized by dividing by $\cos(q_3)$ the control signal. Therefore, the dynamical equation (1) relative to the parameter q_1 along the $\vec{y}$ axis can be written as: $J\ddot{q}_1 + f\dot{q}_1 + kq_1 = -k_u\dot{q}_3 - LF_y$ with $k_u = -B_4\,\Omega$ and $J = J_2 + m_4\left(L_A^2 + L_B^2\right)$. For simplicity, let $d = LF_y$ be the perturbation torque due to the heart. The global control structure is depicted in Fig. 3 where the transfer function $H(s)$ is given by:

$$H\left(s\right) = \frac{k_0 s^2}{s^2 + as + b}$$

with $k_0 = \frac{-L}{J}$, $a = \frac{f}{J}$ and $b = \frac{k}{J}$.

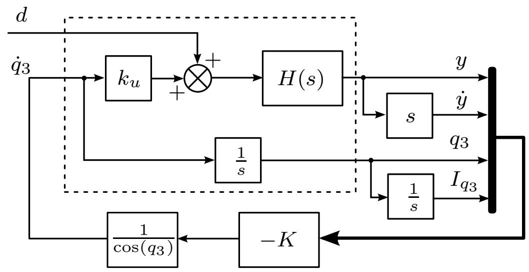


Fig. 3. Structure of the system controlled with static state feedback $K = [k_1\ k_2\ k_3\ k_4]$. Dashed line delimits the gyroscopic system itself.

B. Control issues

The main goal of the control strategy is to reject the effects of the heart beats on the stabilizer movement, i.e. to have the gain of the transfer T_{yd} from heart excitation to position as low as possible within a given bandwidth corresponding to the heart beating frequencies (from 1 Hz to 10 Hz).

The second goal is to keep the position q_3 of the gimbal in the interval $[-q_{3_{\max}}; q_{3_{\max}}]$. Indeed, the gyroscopic torque direction depends on the gimbal angle and would induce for $|q_3| > \frac{\pi}{4}$ a significant lower effect on the desired direction and a disturbance on the other direction i.e. $\vec{y}$ and $\vec{x}$ respectively. This problem can be solved by insuring that the transfer $T_{q_3 d}$ from perturbation to gimbal position is finite in low frequency, thus avoiding angle drift.

Notice that the two aims are conflicting as the perturbation rejection requires the use of the control input. Therefore, it will be necessary to meet a trade-off between perturbation rejection in high frequency and free movement in low frequency.

To satisfy these goals we chose to use the following static state feedback (SSF) control law:

$$\dot{q}_3 = -k_1 y - k_2 \dot{y} - k_3 q_3 - k_4 I_{q_3}$$

where $I_{q_3} = \int_0^t q_3(\tau) d\tau$. The gains k_1 and k_2 allow perturbation rejection on the output y, satisfying the first goal but not the second. Indeed it would induce infinite gain on gimbal position q_3 in low frequency. Including SSF on q_3 allows to make the gain finite, and hence to avoid gimbal drift from an equilibrium position. The SSF on I_{q_3} allows to impose the equilibrium position to zero.

IV. Simulation and experiments

A. Experimental setup

In order to evaluate the active stabilizer, we use the experimental setup depicted in Fig. 4. It is composed of the gyroscopic system attached to a steel rod, which dimensions are similar to those of surgical stabilizers, and a heart simulator. The rod is attached to the table thanks to a tightening device which deliberately presents some compliance to simulate attachement flexibility. The heart simulator is composed of a pan-tilt robot which trajectory is controlled by a sequence reproducing the movements of a heart that were acquired experimentally on a pig [11]. The displacement is converted into force and transmitted to the stabilizer thanks to a compression spring. The gyroscopic system actuators are controlled thanks to a real-time controller (Adept sMI6) which receives position measurement from a potentiometric sensor and computes the control law with a sampling rate of 1 kHz. System monitoring is done from a laptop communicating with the controller trough firewire. It allows to set parameters, launch program sequences and data acquisition. In the sequel the gyroscope speed Ω was set to 30,000 rpm.

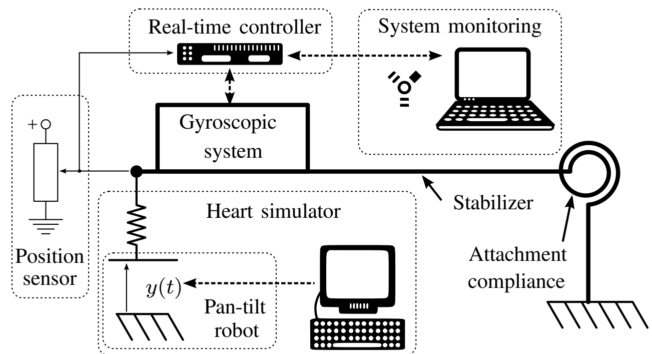


Fig. 4. Description of the experimental set.

B. Identification

To obtain a model which fits optimally the real system, its parameters have been estimated experimentaly. A square signal is sent to the system input $\dot{q}_3$ and the system response y is recorded. Then, the model parameters are tuned using an optimization algorithm that minimizes the least square error between measurement and model. The identification results are illustrated in Fig. 5.

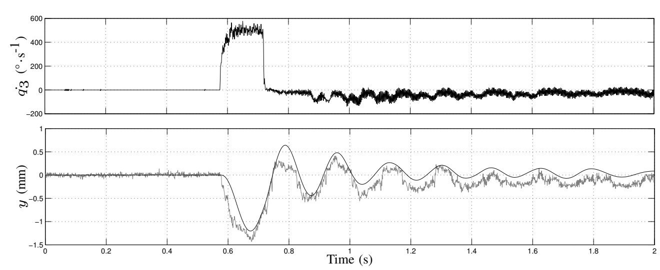
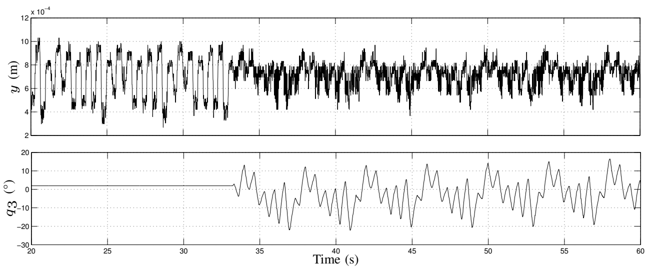


Fig. 5. On top the window control imposed to the system input. On bottom the output response comparison between experimental system (gray) and identified model (black).

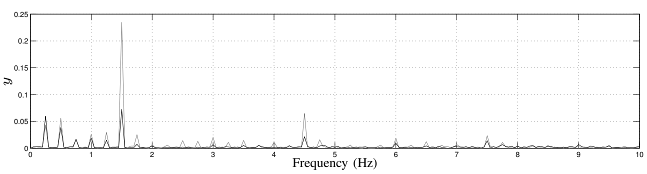
The results exhibit a good fit in terms of amplitude, frequency and damping. However we can observe some slight differences between measured and simulated data. First we can see high frequency components in experimental data which come from measurement noise. We can also notice an offset which is a consequence of a dead-zone in the position sensor. Nevertheless these issues are not significant for the tuning of the controller.

C. First experimental results

The SSF control law has been implemented to control the gyroscopic system. Because of noise and dead-zone on the position measurement, estimation of speed by derivation was problematic. Hence the system has been tested with $k_2 = 0$. Despite this limitation we managed to obtain promising first compensation results which are reported in Fig. 6.



(a) Compensation results for position y and gimbal angle q_3. Compensation is activated at $t = 33$ s.



(b) Comparision of unstabilized (gray) and stabilized (black) results in frequency domains. Data are extracted respectively for $t = 10$ to 30 s and $t = 40$ to 60 s.

Fig. 6. Stabilization experimental results for $K = \{-20000\ 0\ 10\ 0.5\}$.

The compensation effect of the gyroscopic system is visible and we measured a reduction of 48% in the RMS value of the signal. Observing the gimbal angle values reported in Fig. 6(a), one can see that the gimbal angle remains centered around the nominal position. We can also note that since q_3 excursion remains reduced (between -22 and $16°$), undesired displacements along $\vec{x}$ direction remain low. Considering the comparison in frequency domain of measurements with and without compensation which is reported in Fig. 6(b), we can observe that the compensation effect is stronger for high frequencies than for lower. For instance the attenuation of the three first harmonics of heart beating is higher than those of the three first harmonics of breathing.

V. Conclusion and Perspectives

The original concept of gyroscopic actuation to actively compensate for cardiac motion within the framework of minimally invasive beating heart surgery is a promising solution regarding the first experimental results. Thanks to this inertia based actuation principle the system does not need any grounded element, making it fully independant. In addition it is pluggable on commercial instruments without any modification. Thus the whole compensation function is embedded into a rather small and light device easy to manipulate and to integrate in the highly constrained surgical environment.

However, the results we obtained can be largely improved by the use of a more accurate sensing solution. For future work we plan to use acceleration sensing. This solution would be particularly advantageous because it can be fully embedded and does not need any external component linked to the ground since its reference is inertial. In addition it can be fused with position measurement by vision taking advantage of cameras already present in minimally invasive surgery context. Moreover this could allow to let the system work in degraded mode in case of visual obstruction for instance.

VI. Acknowledgements

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References

[1] M. J. Mack, “Minimally invasive cardiac surgery,” *Surgical Endoscopy*, vol. 20, pp. 488–492, 2006.

[2] D. Y. Loisance, K. Nakashima, and M. Kirsch, “Computer-assisted coronary surgery: lessons from an initial experience,” *Interactive CardioVascular and Thoracic Surgery*, vol. 4, pp. 398–401, 2005.

[3] M. Lemma, A. Mangini, A. Redaelli, and F. Acocella, “Do cardiac stabilizers really stabilize? experimental quantitative analysis of mechanical stabilization,” *Interactive CardioVascular and Thoracic Surgery*, vol. 4, pp. 222–226, 2005.

[4] W. Bachta, P. Renaud, E. Laroche, A. Forgione, and J. Gangloff, “Cardiolock: an active cardiac stabilizer, first in vivo experiments using a new robotized device,” *Computer Aided Surgery*, vol. 13(5), pp. 243–254, 2008.

[5] N. C. Townsend, A. J. Murphy, and R. A. Shenoi, “A new active gyrostabiliser system for ride control of marine vehicles,” *Ocean Engineering*, vol. 34, pp. 1607–1617, 2007.

[6] V. J. Lappas, W. H. Steyn, and C. I. Underwood, “Attitude control for small satellites using control moment gyros,” *Acta Astronautica*, vol. 51, pp. 101–111, 2002.

[7] B. Thornton, T. Ura, Y. Nose, and S. Turnock, “Zero-g class underwater robots: Unrestricted attitude control using control moment gyros,” *IEEE Journal of Oceanic Engineering*, vol. 32, pp. 565–583, 2007.

[8] H. Higashimata, M. Yamada, Y. Kazao, and M. Namiki, “Characteristics of active vibration control system using gyro-stabilizer,” *Engineering Structures*, vol. 20, pp. 176–183, 1998.

[9] H. Yano, M. Yoshie, and H. Iwata, “Development of a non-grounded haptic interface using the gyro effect,” in *Proceedings of the11th Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems (HAPTICS03)*, 2003.

[10] P. Cattin, H. Dave, J. Grnenfelder, G. Szekely, M. Turina, and G. Znd, “Trajectory of coronary motion and its significance in robotic motion cancellation,” *Cardio-thoracic Surgery*, vol. 25, pp. 786–790, 2004.

[11] L. Cuvillon, J. Gangloff, M. de Mathelin, and A. Forgione, “Towards robotized beating heart tecabg : assessment of the heart dynamics using high-speed vision,” *Computer Aided Surgery*, vol. 11, no. 5, pp. 267 – 277, September 2006.